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# EVALUATION OF HEAD INJURY IN TERMS OF BIOMECHANICS AND A LOOK AT SPECIFICS OF WHEELCHAIR USERS TRAVELLING IN A CAR

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## SUMMARY

Biomechanics can be generally defined as the application of mechanics to biological systems. Biomechanics of injury then describes the effect of mechanical stress in terms of injury of the human body or human body response to impact and is further exploring the mechanisms by which these injuries occur. Cooperates closely and learns from biomechanics of sports, musculoskeletal system and extreme loads. For orientation in the evaluation of head injury, this study provides an overview of head injury criteria, their mutual comparison and correlation. Problems of wheelchairs is specifies in summary of their passive safety. Wheelchairs are primarily designed for the mobility of the impaired and their passive safety in the event when they will be used as a seat in a car was not taken into account during their development. Effective and efficient restraint systems of the wheelchairs are important for safe transportation of children and adults and the possibility to compare them in terms of injury biomechanics is a highly accented topic.

**Key words:** biomechanics, head injury criterion, impact, passive safety, wheelchair

## INTRODUCTION

### History of injury biomechanics

The history of injury biomechanics began in the nineteenth century when Mr. Messerer dealt with bone strength in bending and pressure. Significant development came with the first and second world war, like in other areas. Into this era, around 1920, falls Hugh DeHavena's research of mechanisms of the human body falling from a height with respect to aircraft crashes. The laboratory research takes place at Wayne State University in 1939 where neurosurgeon Steve Gurdjian and professor of engineering mechanics Herbert Lissner focus on head injury of anesthetized dogs while using skulls of corpses as well. After the Second World War the first mechanical dummies began to emerge. The first mention of mathematical modelling with respect to the computational possibilities of the period also appeared that time. The mechanical dummies began being used for crash tests of cars then and importance of safety features, such as belts, began to grow. A major pioneer of experiments was the U.S. Colonel John Paul Stapp, who in 1955 performed a crash test on himself, when in a rocket traveling at 1000 km/h he slowed to zero velocity in 1.4 seconds. The maximum slowdown reached was 40 G. Subsequently, the biggest propeller of injury biomechanics became a car accident, despite the fact that the first cars appeared already in 1871. The turning point came in 1913, when Henry Ford first started mass production of automobiles, which lowered the price to a quarter and wider public could afford a car. With increasing speed the number of accidents increased as well as their severity and the need arose to create adequate safety features (Hynčík, 2007).

**Biomechanical simulation**

Most knowledge is based on biomechanical simulations. It's hard to determine the exact input factors in real accidents and vice versa it is not easy to simulate a heavy collision with living people in the laboratory.

Back in the early seventies, the biomechanics became interested in of the body damage from mechanical causes. These were mainly injuries in transport accidents and sports. There was an attempt to increase the prevention of such accidents. The main attention was focused on the most common causes of severe head injuries, chest and spine. Based on the known facts there was a mathematical and mechanical elastic-dynamic model developed, which made it possible to monitor the mechanism of head injury during the impact. The model had eleven degrees of freedom and included the skull, brain, spine, upper and lower limbs and trunk mounted to the frame. Further there was examined the behaviour of the scalp, brain and meanings of humans and monkeys and simulated by Maxwell-Kelvin model. Measurements were carried out using implanted accelerometers and pressure meters of human and monkey heads and the results were summarized in a system simulation using a linear impedance model with two degrees of freedom. The issue of injury of spine and its parts was monitored using a model consisting of three-part elements that simulated the entire spine. This simulation was further extended by the mechanical effects of flexors and extensors of the spine.

From the biomechanical point of view we can understand the human body as a closed physical system that can be simulated by a variable spatial system of objects. The structure is composed of subsystems representing different segments of the body. After mastering the given physical situation the interaction of systems as a whole occurs with the external environment (Sychra, 1993).

To describe the injury and its criteria there should be considered an adequate mechanical response of the human body. Variety of tested figurines and various selected collision situations are used and legally supported for this, which have proven mechanical response similar to the human body. These can be examined experimentally or numerically (Hynčík, 2007). They must fulfil the anthropometric aspect (adult, woman, child), testing them must be reproducible with the same results and figure should last many tests. For different types of impacts there are various figurines. Individual dummies can be summarized below:

- Frontal Impact – Hybrid III family, THOR
- Side impact – EuroSID, EuroSID2, SID, SID-hiii, SID-IIs, BioSID, WorldSID

- Rear impact – BioRID, RID2
- Pedestrian – POLAR
- Child – P0, P3 / 4, P3, P6, P10 Q-dummies, CRAB
- Safety belt – TNO-10
- Impactor – head, foot

Standard sizes are 50% male (1.751 m, 78.2 kg), 5% female (1.510 m, 46.82 kg), and 95% male (1.873 m, 102.73 kg).

In the frontal test car the dummy readings in Europe and the U.S. have to meet the criteria:

- FMVSS 208 – HIC <1000, NIJ <1, a3ms <60, longitudinal force in the femur <10 kN,
- ECE R94 – HPC <1000, a3ms <80, VC <1

When the car side tests on dummy readings in Europe and the U.S. meet the criteria:

- FMVSS 214 – TTI <85, apeak <130
- ECE R94 – HPC <1000, VC <1

The numerical models are overtaking great importance in the evaluation of injury criteria as well as injuries themselves in the present. Models can be divided into coupled mechanical systems, the multi-body models (MBS), and detailed models based on finite elements (FE).

There are 4 basic principles of biomechanics of injury: mechanism of injury, mechanical response, tolerance of human body and simulation of impact to the human body (Hynčík, 2007).

**Specifics of wheelchair users travelling in a car**

Wheelchairs are primarily designed for the mobility of the impaired and their passive safety in the event when they will be used as a seat in a car was not taken into account during their development (Dsouza & Bertocci, 2010). 70% of wheelchair users are driving their private vans and are using wheelchair as a separate place to sit with standard restraint systems (VanRoosmalen, Bertocci, Ha, & Karg, 2001). Effective and efficient restraint systems of wheelchairs are important for safe transportation of children and adults who remain seated in their wheelchairs when travelling in motor vehicles, but more emphasis should be placed on proper use of safety belts. This means that manufacturers of wheelchairs must pay attention to the possibility of proper use and placement of seat belts for both passengers and drivers in the design (Schneider, Klinich, Moore, & MacWilliams, 2010). As a result, the people seated in a wheelchair are in a higher risk of injury than those seated in the classic car seat in an accident. The results of studies concerning the transport of disabled children show that a high percentage of such children who are transported daily suffer from poor control of the position of the head and torso, and therefore cannot sit upright without support (Everly et al., 1993). Disabled children who must travel in their chairs are not included in the regulations and laws for the safety acceptance of the cars (Ha & Bertocci, 2007). Not only children but also disabled adults very often sit in their chairs in the public or private transport (Ha & Bertocci, 2007).

Previous automotive safety studies have shown that the integrated restraint systems provide excellent impact protection (Bertocci & Evans, 2000). The study assessed the operational benefits and effectiveness of protection against impact with integrated restraint systems when using the wheelchair as a separate seat. Through a comprehensive comparative analysis of the risks of injury associated with the different settings of restraint systems was found that the integrated restraint system provides effective protection of the person in a wheelchair during impact (Bertocci & Evans, 2000). The main advantage of using integrated restraint system is the possibility of optimal positioning of safety belts. The position of the seat belt over the wheelchair seat is inappropriate when using standard belts. A study by (VanRoosmalen et al., 2001) compares the safety of FWORS (fixed vehicle mounted wheelchair occupant restraint systems) and WIRS (wheelchair integrated restraint system) in the sledge test with load of 20 G in frontal impact at 30 mph and proves that the most important for protection during frontal impact is the correct positioning of harnesses of the upper body, especially for people with atypical sitting heights. Because the persons sitting in a wheelchair are not usually able to use standard seat belts mounted by a manufacturer and because the seats are an important part of passenger protection during impact, it is clear that a threat of serious and fatal injuries is much greater for them than for those who sit in a standard seat with a safety belt (Schneider et al., 2010).

In response to this problem new standards were adopted for wheelchairs used as a car seat (ANSI / RESNA WC19). These standards provide general design requirements and test methods for wheelchairs used as seats in motor vehicles. WC19 using 20G load at a speed of 48 km/h (30 mph) in the frontal impact during the assessment of dynamic rigidity and evaluation of load during an impact in the driving direction of a wheelchair. These tests thus provide a view on the wheelchair users loading and assess the resistance of wheelchairs to impact, which helps their development. Dynamic tests require the use of expensive equipment and complex measurements, therefore computer simulations are used as a parallel approach that can provide economic and versatile method of analysis of accidents with wheelchairs. Nevertheless the model is only an approximate representation of real world and needs the most accurate validation. Validation is used to determine how accurately the model represents the real world and can be imagined as a calibration tool. Validation consists of repetitive processes of comparing model results with experimental results. The model is then tuned to minimize differences between experimental systems and results from the model.

When comparing the Dsouza and Bertocci (2010) model with other computer simulations it was found that most studies used peak acceleration values between the models and sledge tests as the primary evaluation criteria. Kang and Pilkey (1998) developed a model of wheelchair for the front impacts, in which they compared the maximum percentage difference during the validation. The peak values were found to vary from 0% to 30%. According to a study of Bertocci, Szobota, Hobson, and Digges (1999) a model DynamanTM was developed and validated from 50% model of the adult male on an electric wheelchair in a frontal collision. Validation of the model was made by comparing the model and sledge test during movement between the wheelchair user and the wheelchair. Peak value varied from 1% to 25%. Ha and Bertocci (2007) developed a manual wheelchair model occupied with a child dummy Hybrid III (6 years), which was verified with the value of the percentage difference between 2% to 16% with the Pearson correlation coefficient having an average evaluation criterion 0.91.

To improve the security of wheelchair users in transport there were national and international organizations J2249 *Wheelchair Tiedowns and Occupant Restraint Systems* (WTORS) and International Standards Organization (ISO) 10542 *Wheelchair*

*Tiedown and Occupant Restraint Systems* established that focus on specific needs of wheelchair users, test methods and the development of security systems (Ha & Bertocci, 2007). In the absence of detailed knowledge about the biomechanical response of children to impact, load values were converted from 50% adult men, thus creating the injury criteria for 6HybridIII, a six years old boy dummy (Hybrid III 6-year-old – 6HybridIII) which was developed and have been used for experiments to improve the security of wheelchair users during a frontal collision in a car (Ha & Bertocci, 2007), which according to FMVSS 208 include HIC 15 chest acceleration, chest compression, Neck Indry criterion, peak tension and compression level of the neck and according to FMVSS 213 include HIC without limitation of time interval, HIC 36, peak acceleration of the chest. For HIC 15 there was established a limit of 700, for the HIC 36 a limit of 1000 (Ha & Bertocci, 2007).

For specific safety assessment the injury criteria CIC and MC were created. Combined injury criterion (CIC) reflects the risk of injury associated with trauma of the important body part. Quantitative value of injury of each body part was determined according to accident statistics from the U.S. and Sweden (Viano & Arepally, 1990). Individual injury ratios according to importance are: head 47%, neck 12%, chest 23% and abdomen 18% (VanRoosmalen et al., 2001). High level of injuries indicates that injuries associated with this body part will be the most serious and may thus be the cause of death.

$$CIC = 0.53\left(\frac{HIC}{1000}\right)_{head} + \frac{0.18}{4}\left(\frac{F_{tens}}{247} + \frac{F_{comp}}{247} + \frac{F_{shear}}{247} + \frac{M_{flex}}{1681}\right)_{neck} + 0.29\left(\frac{a}{60}\right)_{chest} \qquad \text{[Eq. 1]}$$

where HIC is a head injury criterion, $F_{\mathrm{tens}}$ is axial loading - neck tension, $F_{\mathrm{comp}}$ is the axial loading – neck compression, $F_{\mathrm{shear}}$ is the neck shear load, $M_{\mathrm{flex}}$ is the neck torque load and *a* chest is the chest acceleration.

Motion criterion (MC) is based on the principle of momentum and weight and a presents a risk associated with contact with the interior element. MC index can be used to compare kinematic responses to various restraint systems.

$$MC = 0.25\left(\frac{Exchead}{25.6}\right) + 0.25\left(\frac{Excknee}{14.8}\right) + 0.25\left(\frac{Excwc}{7.9}\right) + 0.25\left(\frac{\frac{Excwc}{Excknee}}{1.1}\right) \qquad \text{[Eq. 2]}$$

where *Exchead* is a head track, *Excknee* is a knee track, *Excwc* is a wheelchair track and *Excwc* / *Excknee* is the ratio of wheelchair and knee tracks. MC and CIC indices can range from 0 to 1, where higher values mean a higher risk of injury.

## MATERIAL AND METHODS

### Wayne State Tolerance Curve

The first extensive evaluation of head resistance to impact was the Wayne State Tolerance Curve (WSTC). WSTC shows that when exposed to linearly accelerated violence in non-breach load both the size and duration of acceleration pulse have influence on the risk of brain injury. Short duration, high acceleration impact (2ms and 400 g) leads to the same risk of injury as a long-lasting impact and low acceleration (30ms and 80g). WSTC served as the basis for injury criteria used in car crash regulations.

**Acceleration Severity Index**

Another criterion assessing the severity of injuries is Acceleration Severity Index (ASI). It is defined as

$$ASI(t) = \left[\left(\frac{\bar{a}_x}{\hat{a}_x}\right) + \left(\frac{\bar{a}_y}{\hat{a}_y}\right)^2 + \left(\frac{\bar{a}_z}{\hat{a}_z}\right)^2\right]^{\frac{1}{2}} \qquad \text{[Eq. 3]}$$

where $\hat{a}_x$, $\hat{a}_y$, $\hat{a}_z$ limit values for acceleration along the x, y, z axes and $\bar{a}_x$, $\bar{a}_y$, $\bar{a}_z$ are acceleration values of the selected point in the vehicle. This point is usually placed in the front seat on the driver's seat.

$$\begin{aligned} \bar{a}_x &= \frac{1}{\delta}\int_t^{t+\delta} a_x dt; \\ \bar{a}_y &= \frac{1}{\delta}\int_t^{t+\delta} a_y dt; \\ \bar{a}_z &= \frac{1}{\delta}\int_t^{t+\delta} a_z dt \end{aligned} \qquad \text{[Eq. 4]}$$

The average time interval was set at Acceleration limit values are set so that passengers would face little or no risk of injury AIS < 2. For the passenger strapped by a seat belt the following values apply:

$$\hat{a}_x = 12 \text{ g}; \hat{a}_y = 9 \text{ g}; \hat{a}_z = 10 \text{ g} \qquad \text{[Eq. 5]}$$

**Head Injury Criterion**

To assess the severity of injury was used Head Injury Criterion (HIC), which is defined as:

$$HIC = \left\{(t_2 - t_1)\left(\frac{1}{t_2 - t_1}\int_{t_1}^{t_2} a(t)dt\right)^{2.5}\right\}_{\max} \qquad \text{[Eq. 6]}$$

where *a(t)* is the resultant acceleration of the head and $t_1$ and $t_2$ are variable initial and final time intervals during which HIC reaches its maximum value. For regulation purposes the maximum interval between $t_1$ and $t_2$ set at 15, or 36 ms. Head injury criterion HIC use is based on proposal of the National Highway Traffic Safety Administration (NHTSA), 1972 (Marjoux, Baumgartner, Deck, & Willinger, 2008).

For the effects of direct impact has been demonstrated that HIC is acceptable discriminator between severe and less severe injuries (Tarriere, 1981). It also correlates with the risk of fractures of the skull (Ran, Koch, & Mellander, 1984). However, for the shocks from different directions a poor correlation was found between HIC and seriousness of injury, because the rotation of the head is not taken into account, often being the primary cause of various types of traumatic brain injury (Marjoux et al., 2008). HIC taking head rotation into account has also been suggested but never thoroughly evaluated (Brands, 2002). HIC predicts the risk of injury from external mechanical impact to the head, which can be measured directly from the crash test dummy, but does not take into account the internal mechanical response. Furthermore, there is no distinction between different types of traumatic brain injury. For research on the so-called "next generation wound" a computational model head was used. More detailed description of the injury was achieved by using the calculated internal mechanical response, resulting from external mechanical impact to the dummy (Brands, 2002). Examples of such injury determination are SIMon, a simulated injury monitor (Bandak et al., 2001), Head Impact Power (HIP) or GAMBIT (Newman, Shewchenko, & Welbourne, 2000). This injury criterion combines the sliding and rotational acceleration. Given the assumption that the head is just as prone to translational as rotational acceleration, then the value 1 is taken as the limit.

$$GAMBIT = \left[ \left( \frac{a(t)^n}{a_c} \right) + \left( \frac{\ddot{\varphi}(t)^m}{\ddot{\varphi}_c} \right) \right]^{\frac{}{k}} \text{; where m = n = k = 2.5} \qquad \text{[Eq. 7]}$$

Determination of time interval appears to be an important parameter for calculating the HIC value. Length of time interval is set to t = 36 ms and for the analysis of the hard impact of the head to t = 15 ms. According to EHK 94 regulation (First, 2008) the threshold is then determined as HIC = 1000 and the acceleration that is greater than 80 g for no longer than 3 ms. $HIC_{15}$ value may be directly correlated with the $HIC_{36}$ value by the formula $HIC_{15} = 0.7 \times HIC_{36}$ . $HIC_{36}$ was designed to protect the head against injuries such as fractures of the skull with a longer exposure, when there is no contact with the hard parts of the interior. $HIC_{15}$ was calculated from the short duration impact on a hard and heavy surface from deadly head impact data and was designed to minimize both the fracture of the skull and brain injury caused by head contact with the interior equipment. Short duration impact may include a direct impact of driver's head to the edge or centre of the steering wheel or the child's head impacting to the hard part of the dashboard.

## Severity of injury

Since it is not possible to describe in detail a wide range of injuries which may occur, there was an AIS (Abbreviated Injury Scale) scale derived in 1977, which describes the injury values from zero to six. This scale is continuously updated and each body part is covered by a different description. The scale is based on a healthy adult, so it is necessary to take the criteria into account with regard to other traffic participants. Zero means no injuries and six means injuries incompatible with life. The scale is not linear. Since the scale is different for different parts of the body, there is a value MAIS, which means the maximum AIS and contains the maximum for all parts of the body. Injuries severer than e.g. AIS3 are then indicated AIS3+. In addition to the AIS scale, which actually describes the injury only immediately after the actual incident (Hynčík, 2007), there are other scales. ISS (Injury

Severity Score) divides the body to the head and neck, face, chest, abdomen, pelvis and limbs, including the outer surface (which issues such as scratches and burns are related to). ISS is the sum of all peaks of AIS on the body and is in the range from 0 to 75. If AIS6 appears on some body part, ISS automatically equals 6 too. Equally important is the economic scale, which is trying to relate and long-term development of the patient with regard to the cost of treatment, rehabilitation and work disability. Let us mention the ICS scale (Injury Cost Scale) here. Physical parameters involved in the scales are the most easily measurable quantities such as acceleration, velocity, or other derived quantities such as energy equivalent speed (EES) or coefficient of restitution, which determines the elastic and plastic strain ratio.

AIS table (Table 1) is focused on head injuries and divides the severity of injuries according to the value of AIS to grade of brain concussion, also provides clinical and pathological description of injury and injury effect after one month (Ommaya, Thibault, & Bandaki, 1994).

**Table 1.** Severity of injuries according to the value of AIS (Ommaya et al., 1994)

| AIS level | Concussive brain injury grade | Clinical desriptions | Pathologic desription | Outcome (1 month) |
|---|---|---|---|---|
| 1 | I | “Ding”, “Stunned without amnesia”. Minor symptoms e.g. headaches and ataxia | Not known; CT and MRI scans usually normal; skull fractures and intracranial bleeding uncommon. PET scans may be abnormal. | Normal except with a TADD syndrome or a post-concussive syndrome develops |
| 2 | II | Amnesia without coma (type A, slow onset; type B, immediate onset) | | |
| 3 | III | Coma <6 h. (includes classic cerebral concussion, minor and moderate head injuries) | Increasing intensity and distribution of diffuse lesions and/or intracranial bleeding (e.g. Acute subdural clots); CT and MRI scans usually abnormal; skull fracture 20–50% | |
| 4 | IV | Coma 6–24 h. (severe head injuries) | | Morbidity increasing to +35% and mortality to 50% |
| 5 | V | Coma <24 h. (severe head injuries) | | |
| 6 | VI | Coma /death within 24 hours (fatal head injuries) | | |

RESULTS

## Comparison of HIC and AIS

Based on HIC values and AIS tables we can estimate the risk of brain damage but not damage to the facial part of the skull. The common head injury in a traffic accident include the facial area injury especially from hitting the steering wheel, dashboard, etc., which may be associated with brain contusion. From the injuries in accidents can be concluded that the direct impact of face to the steering wheel or dashboard is much more dangerous than the areal impact of face into the airbag. Attempts have been made with measuring foils applied to the dummy human face. When evaluating the experiment it was shown that HIC is roughly the same, but the areal compression is much lower (Sychra, 1993). It means that the areal compression at the point of contact is crucial to the injury.

Based on experiments on dead bodies a comparison was determined between $HIC_{36}$ value and AIS table (Shojaati, 2003).

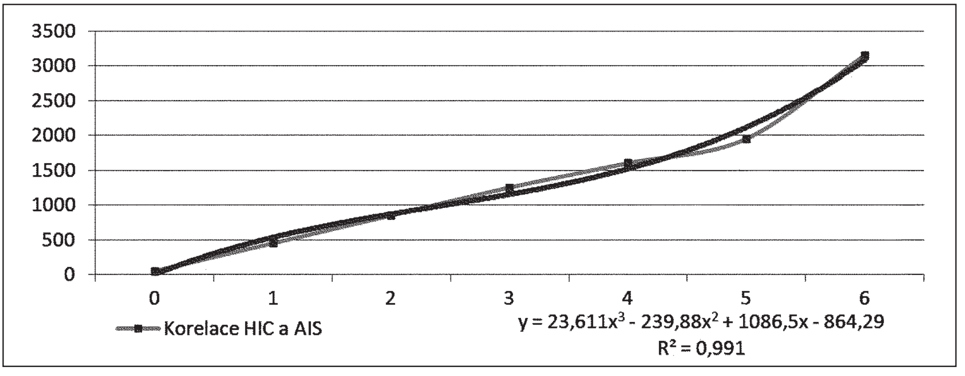


**Figure 1.** Comparison of $HIC_{15}$ with table AIS

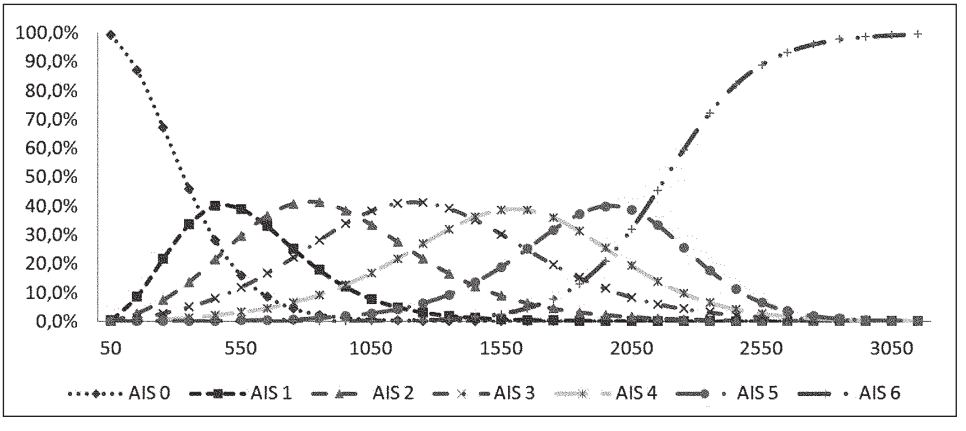


**Figure 2.** Relative frequencies of injury severity according to the value of $HIC_{15}$

Based on Table 4 of relative frequencies of injury severity according to the value of $HIC_{15}$ published by Prasad and Mertz (1997) I have set a graph (Figure 1) comparison value of $HIC_{15}$ and injury severity (AIS) and graphically illustrating the relative frequency of injury severity (Figure 2 ) by the value of $HIC_{15}$.

**Comparison of EES and HIC**

According to the update of table of relative frequencies of injury severity to match the severity of injury graph of 219 strapped crew members from the number 7 values of AIS from 0 to 6 to 4 values of 0.1, 2, 3, 4–6, I set the graph of the relative frequency of injury severity according to updated AIS (Figure 3).

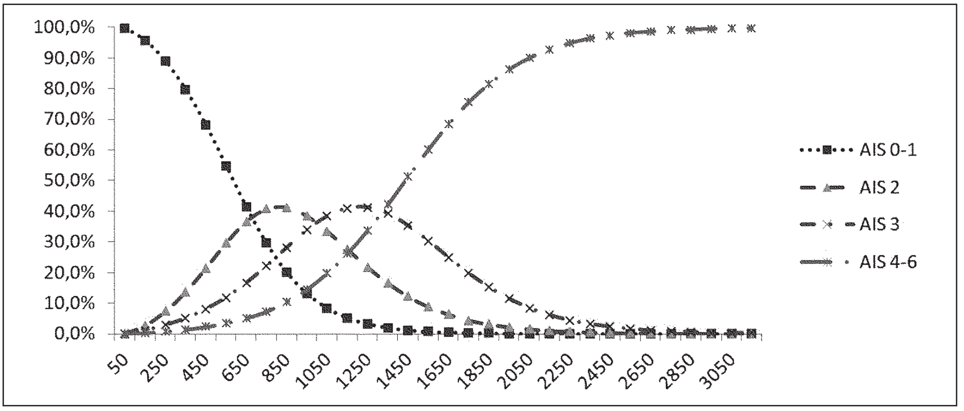


**Figure 3.** Relative frequency of injury severity according to updated AIS

Now we can interpolate to determine the HIC value appropriate to EES of 219 strapped crew members (Figure 4).

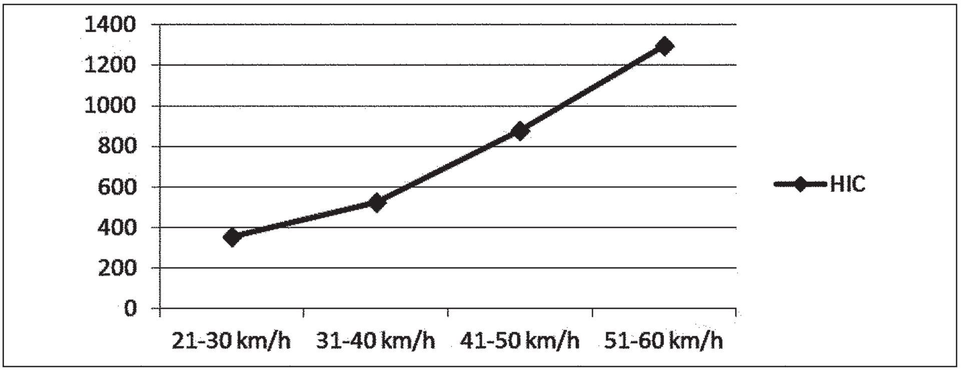


**Figure 4.** EES and HIC strapped

Similarly, we interpolate values of 133 non-strapped crew members (Figure 5).

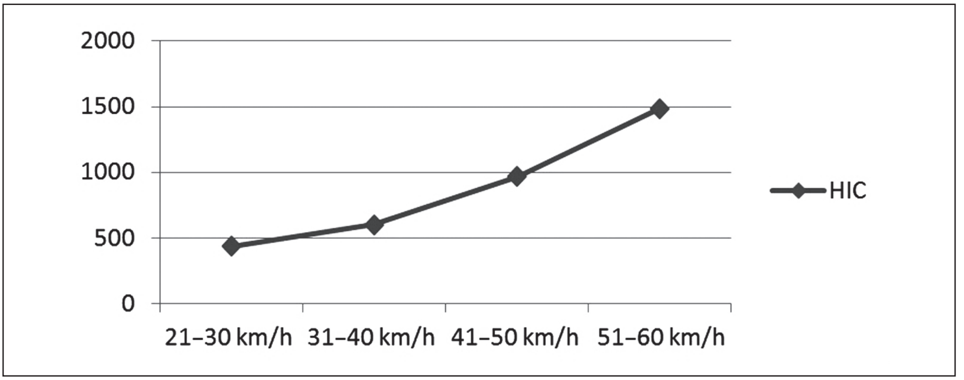


**Figure 5.** EES and HIC non-strapped

We can compare the increase in head injury criterion between strapped and non-strapped crew members.

**Table 2.** Comparsion of strapped and non-strapped crew

| EES | HIC strapped crew | HIC non-strapped crew |
|---|---|---|
| 21–30 | 353 | 435 |
| 31–40 | 522 | 601 |
| 41–50 | 878 | 968 |
| 51–60 | 1297 | 1488 |

## Multibody experiment simulation – biomechanical response of the head on impact

In order to validate a model and compare biomechanical response of living head to impact there was a comparative simulation carried out using a MADYMO software (MADYMO, 2009). The model of 50% male was used from the database program for this purpose. He was seated in a position and there was an impactor model created corresponding to experiment. For more see (Fanta, Kubový, & Jelen, 2010).

**Table 6.** Comparison of real impact and multibody simulation

| HIC | Tested O | Tested P | Multibody simulation |
|---|---|---|---|
| **/001** | 19 | 18 | 23 |
| **/002** | 97 | 73 | 101 |
| **/003** | 188 | 177 | 195 |
| **/004** | 36 | 27 | 38 |
| **/005** | 128 | 110 | 145 |
| **/006** | 215 | 185 | 239 |

**Figure 6.** See Colour Appendix

After the simulation can be stated that the model from the MADYMO database shows the same biomechanical head response to impact and the value of head injury shows identical results as the real experiment and the model is therefore suitable for further use in the assessment of head injury.

## DISCUSSION

Biomechanics of head injury is a complex system that includes knowledge of the anatomy, traumatology, mechanics, and not least the knowledge of mathematical modeling. Issue of wheelchairs is another possible direction for the biomechanics of injury. According to the research literature, testing at the level of mathematic simulations using multi body systems and crash tests are under way. Crash tests require that the wheelchair, including the wheelchair frame and seating system, be sled-impact tested using a 20 g/48 km/h frontal crash pulse. Making experiments is very expensive process and this is the reason, why using computer simulations are increasingly popular.

However for proper implementation of the model must be known the actual response of the organism and probability of injury. For this purpose it was created comparison of HIC and AIS and then was assigned to the energy equivalent speed, which is used to roughly determine the extent of injury based on the input speed. It is likely that a given problem cannot be completely accurate, because of large variability in initial conditions. For higher accuracy, it would be necessary to model the situation in the finite element solver.

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## HODNOCENÍ PORANĚNÍ HLAVY Z HLEDISKA BIOMECHANIKY A POHLED NA SPECIFIKA VOZÍČKÁŘŮ CESTUJÍCÍCH V OSOBNÍM AUTOMOBILU

SOUHRN

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Biomechaniku můžeme obecně definovat jako aplikaci mechaniky na biologické systémy. Biomechanika poranění pak popisuje efekt mechanického zatížení ve smyslu poranění lidského těla nebo odezvy lidského těla na náraz a dále se zabývá i mechanismy, kterými tato poranění vznikají. Úzce spolupracuje a čerpá poznatky z biomechaniky sportu, pohybového systému a extrémní zátěže. Pro orientaci v hodnocení poranění hlavy přináší tato studie přehled používaných kriterií poranění hlavy a jejich vzájemné porovnání. Problematika vozíčkářů je pak specifikována v přehledu jejich pasivní bezpečnosti. Invalidní vozíky jsou primárně určeny pro mobilitu postižených a při jejich vývoji nebyl brán zřetel na pasivní bezpečnost v případě, že budou použity jako samostatné místo k sezení v osobním automobilu. Efektivní a účinné zadržovací systémy vozíků jsou důležité pro bezpečnou přepravu dětí a dospělých a možnost jejich srovnání z pohledu biomechaniky poranění je vysoce akcentovanou problematikou.

**Klíčová slova:** biomechanika, HIC, náraz, pasivní bezpečnost, invalidní vozík

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